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EVIDENCE FOR THE UTILIZATION OF DYNAMIC PRELOAD IN IMPACT INJURY PREVENTION

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SUMMARY

Dynamic preload is anticipatory acceleration in the same direction as a later impact acceleration. To evaluate the influence of dynamic preload on human impact response, tests with volunteer subjects were conducted on impact facilities at the Air Force Aerospace Medical Research Laboratory (AFAMRL). Test data are presented which indicate that the peak forces and body segment accelerations imposed on subjects during impact accelerations are decreased when those impacts are preceded by dynamic preload. The impact response differences were more striking for comparisons between zero and low levels of dynamic preload than for comparisons between low and higher levels of preload. The threshold for these protective effects is apparently below 0.25 G dynamic preload for the test conditions investigated. In addition, the medical and subjective data support the assertion that dynamic preload is protective when applied prior to $-G_x$ impact accelerations. Since impacts conducted on decelerator facilities are all influenced by track friction and therefore preceded by dynamic preload, it appears that they are fundamentally different from impacts conducted on accelerator facilities, involving zero dynamic preload. This indicates a need to reassess previous tolerance estimates derived from rocket sled decelerations. Decelerator tests do not appear to predict the more severe results of similar exposures on accelerators. Research efforts are continuing at AFAMRL to further delineate the significance and utility of dynamic preload as a technique in impact injury prevention.

INTRODUCTION

A number of well recognized factors influence human response to impact acceleration. These include differences in restraint harness materials, geometry, and pretension; biological variability among subjects; and variations in subject posture and voluntary bracing. A less well recognized factor of fundamental importance in human impact tolerance appears to be the acceleration-time history imposed on the subject prior to the impact event. Interest in this pre-event history has led to the concept of dynamic preload.

Dynamic preload is defined as an imposed acceleration preceding, continuous with, and in the same direction as an impact acceleration pulse. Some impact accelerations, such as those of a body at rest or a body moving at constant velocity, occur with zero dynamic preload. The pre-impact acceleration in these cases is zero. Other impact accelerations, such as those of a body moving at a decreasing velocity, occur with a variable amount of dynamic preload. Such would be the case in a moving automobile with the brakes applied before striking a barrier.

Dynamic preload should not be confused with static preload, as might be applied by pretensioning a harness restraint system. Static preload has been demonstrated to be useful, and some of the effects are interrelated, but dynamic preload produces additional effects not attainable through the use of harness tension. Dynamic preload should also not be confused with voluntary subject bracing (such as with the extremities) or subject pre-positioning (such as flexing the head forward prior to a $-G_x$ impact). These techniques have also been demonstrated to influence human impact response. However, unlike dynamic preload, to effectively influence response, bracing and pre-positioning require not only subject anticipation of the impact, but also proper subject performance of the technique as well.

In the conduct of experimental impact testing, facilities have been used which vary in the dynamic preload they impose. Data gathered in the attempt to explore human tolerance limits are derived primarily from experiments conducted on decelerator facilities. The early impact sled tests used rocket thrust to accelerate a sled to a desired velocity (4, 8). The sled then coasted into some form of mechanical or hydraulic braking device which applied the retarding force necessary for the planned impact. During the coast phase, however, retarding forces were already at work in the form of wind resistance and rail friction. These pre-impact retarding forces generally produced sled accelerations in excess of 1.5 G's, sometimes reaching 15 G's. (See Table 1.) The higher levels of imposed dynamic preload often produced dramatic impact responses in the human occupants of the sled well before contact with the brake was made. One of John Paul Stapp's rocket sled exposures, for example, was described as follows.

"At burnout of the rockets, the subject's head and shoulders were pitched forward abruptly into the harness and firmly pressed against the straps throughout the 1.6 seconds of coasting."

The acceleration-time curve for this run demonstrated a 4 to 5 G deceleration of the test sled during the coast phase. Although Lombard (4) suggested over 15 years ago that the timing of this "double punch" profile could be of analytical interest and importance, the protective implications of these data were only recently suggested by Raddin et al. (7).

TABLE 1. DYNAMIC PRELOAD AND CORRESPONDING MAXIMUM ACCELERATION TOLERATED DURING $-G_X$ HUMAN IMPACTS AT VARIOUS TEST FACILITIES

TEST FACILITY	RANGE OF DYNAMIC PRELOADS (G)	MAXIMUM TEST LEVEL (PEAK G)
Rocket Sled Decelerator	1.5 (Calculated Average) to 16.8 (Measured Peak)	45.5 (Measured at seat)
Daisy Decelerator	0.2 - 0.3	34.4 (Measured at sled)†
Accelerators	0	15-16

† Subject incurred multiple vertebral fractures and shock as the result of this exposure.

Other decelerator facilities, such as the Daisy Decelerator at Holloman Air Force Base and the Horizontal Decelerator at AFAMRL impose less dynamic preload on subjects than that experienced during the early rocket sled experiments. The $-G_X$ impact tolerance work conducted on the Daisy Decelerator (1) resulted in a lower maximum test level than the previous experiments. As shown in Table 1, this exposure level unfortunately resulted in untoward medical effects and thus represented a level of objective tolerance for those test conditions.

Impact facilities designed to produce impact accelerations from a standing start always impose zero dynamic preload. The maximum test levels achieved on such accelerator facilities have been well below the levels achieved on decelerator facilities. In one series of $-G_X$ experiments conducted on an accelerator (5), the highest level of human exposure reported was 15 G (57 ft/sec). Therefore, as shown in Table 1, previous experience in $-G_X$ impacts suggests that human tolerance to impact increases with increasing dynamic preload. Although the impact acceleration-time profile for two facilities may be identical, the acceleration-time history prior to the event may differ greatly and may be the basis for differences observed in impact response.

It appears that significant protective benefits accrue to the subject of a test with imposed dynamic preload. At very high levels of preload, body segments are rearranged in advance of the impact, preventing the amplified accelerations associated with the head or other body segment snapping forward during the impact. In effect, the segment moves forward first to avoid snapping forward later, and experiences lower forces during the deceleration. In a similar manner, each small volume of body tissue can be considered to act in some ways like a supported segment. Each volume can be considered to have an associated dynamic preloading threshold which, when applied, can serve to minimize its later impact response. In this way a low level of dynamic preload can be seen to potentially produce beneficial results without associated motion of large body segments or involvement of voluntary bracing.

TABLE 2. NUMBER OF INJURIES INCURRED BY BABOON CADAVERS DURING $-G_X$ IMPACTS (50 G)

TEST FACILITY	ACCELERATOR	DECELERATOR
Subjects	6	6
Significantly Injured Subjects	4	1
Fractures	1	1
Muscle Tears	4	0
Liver Tears	1	0

Similar apparent differences in tolerance, injuries, and response accelerations have been noted with animal surrogates. For example, 12 baboon cadavers were exposed to nominal 50 G impact accelerations at AFAMRL. As shown in Table 2, of the six subjects exposed on the accelerator, four incurred significant injuries, including one clavicular fracture, one hepatic laceration, and four transections of the rectus abdominis muscles. On the other hand, of the six baboon cadaver subjects exposed to a comparable impact on the decelerator, only one incurred a significant injury. These differences may also be ascribed to dynamic preload.

Comparison of data derived from different impact test facilities, subjects, and conditions has always been difficult. Similar peak G exposures may have very different pulse shapes and, therefore, different velocity changes and pulse energy content. Restraint system designs and materials are often not comparable. Subjects differ in size, weight, and response characteristics. In short, if clear distinctions between responses with differing dynamic preloads are to be made, great care must be taken to

The effect of varying the acceleration-time history preceding the impact event while carefully controlling all other sources of response variance is the subject of the present inquiry. The results of matched impact exposures on a decelerator (preload levels of approximately 0.25 G and 0.62 G) and an accelerator (no preload) are presented. The study is also an attempt to establish deceleration exposures as a special case in human tolerance data. Presumed tolerance information derived from tests with high preload would then require interpretation when applied to impact tests with no preload. Tolerance scaling techniques would be required. Furthermore, if tolerance increases with preload, means should be sought to intentionally impose preload as a protective technique in impact exposures.

EXPERIMENTAL DESIGN

This experiment was designed to provide a controlled comparison of human response in matched impact acceleration profiles on decelerator and accelerator facilities. The comparable tests were matched for velocity change of the impact sled. The impact acceleration profiles were approximate half-sine waveforms. The tests were conducted on the Horizontal Decelerator and the Impulse Accelerator at AFAMRL. The ready availability of these facilities for use in human testing and the extensive base of comparative test data were prime factors in selecting forward facing ($-G_x$) tests for this investigation.

Volunteer subjects came from the AFAMRL Impact Acceleration Stress Panel. Prior to participation, all subjects successfully completed a thorough medical screening evaluation, including a USAF Flying Class II physical examination, pulmonary function tests, electroencephalogram, exercise treadmill test, and a complete battery of skull, chest and spine x-rays. This screening procedure has been more thoroughly described elsewhere (3). Ongoing informed consent was provided by all subjects throughout the experiment in accordance with the applicable human use guidelines as defined in Air Force Regulation 169-3.

To minimize the potential for injury to subjects, the tests were conducted at presumed subinjury impact acceleration levels. The experimental design matrix is shown in Table 3. The pulses in Test Conditions B and E were preceded by the minimum dynamic preload (approximately 0.25 G) created primarily by friction between the impact sled and the track rails of the decelerator. In Test Conditions A and D, an additional approximate 0.35 G dynamic preload was imposed by the application of sled-mounted braking devices approximately 250 msec prior to impact. There was no dynamic preload in Test Conditions C and F, which were conducted on the accelerator. The 8 G sled profiles are shown in Figure 1 and the 10 G profiles are shown in Figure 2. The forces acting on the subjects at these exposure levels are generally sufficient to overcome the forces of voluntary muscle contraction and, therefore, produce a response which is suitable for comparative parametric analysis. The preload conditions, however, were not sufficient to cause an observable change in subject posture prior to initiation of the impact.

TABLE 3. NOMINAL TEST CONDITIONS

TEST FACILITY	DYNAMIC PRELOAD (G)	CONDITION	
Decelerator	0.62	A	D
Decelerator	0.25	B	E
Accelerator	0	C	F
Sled Acceleration (G)		8	10
Sled Velocity Change (ft/sec)		27	30

The test seat was designed with conventional USAF crew seat geometry with a seat back angle of 13° aft of vertical and a seat pan inclined 6° above the horizontal. Leg bracing was not possible since no footrest was provided. The subjects were restrained by a lap belt and double shoulder strap harness constructed of 1 3/4 inch wide webbing. The straps were pretensioned to 20 ± 5 pounds. Prior to each impact test, the subject was instructed to assume the same body posture, with head against the headrest, hands resting on anterior thighs without upper extremity bracing, and posterior thighs in contact with the seat pan. During exposures on the accelerator, the subject could hear the countdown to impact in order to assure cognitive anticipation of the event similar to that achieved during decelerator tests.

The test fixture, restraint harness, and subject were instrumented to obtain pertinent objective data during each test. Measured parameters included impact acceleration of the test sled and seat, impact velocity of the test sled, loads reacted at the seat pan, and loads measured at the restraint harness attachment points. Accelerations at the head and chest of the subject were measured by appropriately mounted triaxial translational accelerometers. Photometric data were obtained by two high-speed (500 frames per second) motion picture cameras mounted on the test fixture, permitting assessment of body segment displacements during the impact.

Subjective data were also obtained by means of a post-test questionnaire designed to assess the subject's impression of each impact event relative to other comparable exposures. For example, subjects were asked to characterize head displacement as small or large, shoulder strap pressure as low or high, and overall impression of the impact as comfortable or uncomfortable on an integer scale from -3 to +3, zero indicating a neutral

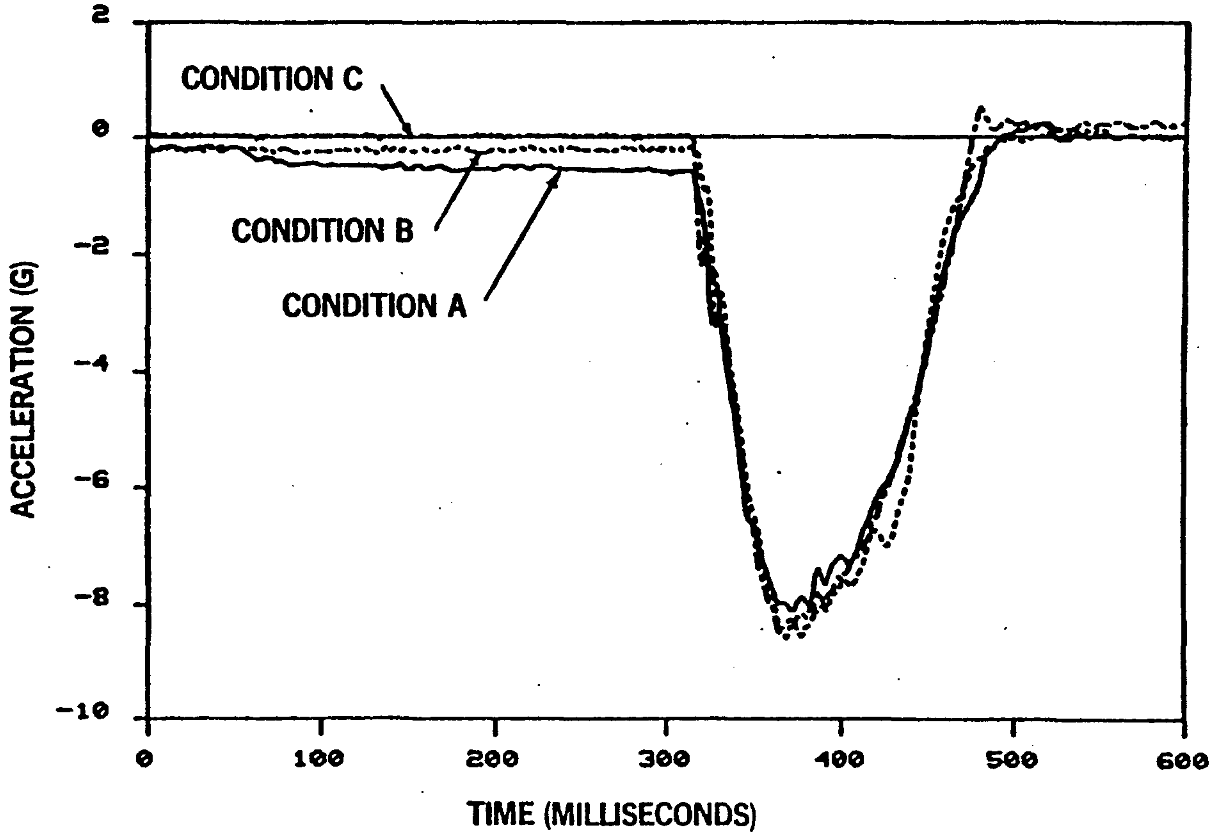

Figure 1. Typical Sled Acceleration Profiles at the 8 G Test Level.

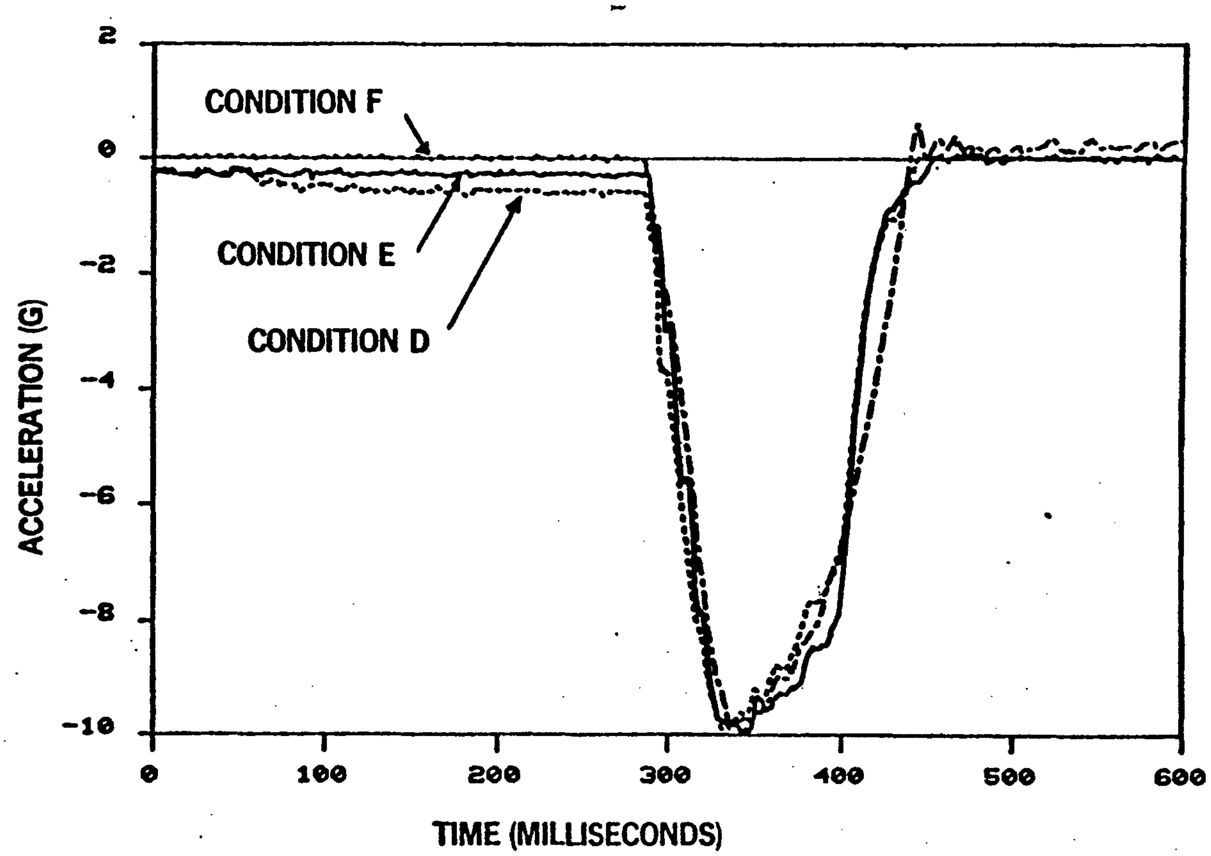

Figure 2. Typical Sled Acceleration Profiles at the 10 G Test Level.

reaction. Subjects were not permitted to review previous post-test responses when completing the questionnaires.

The electronic data were processed by computer and the test results were evaluated using the Wilcoxon paired-replicate rank test (9). This statistical technique was selected to compare the peak values of specific measured parameters and to establish the statistical significance of observed trends in the data. Experimentally measured parameters for each subject were arithmetically compared with the same parameter measured for the same subject in a comparable test condition, in order to establish pair differences. When a sufficient preponderance of ranked pair differences for a parameter changed in the same direction, a trend was established as statistically significant by the Wilcoxon analysis. The 90% confidence level (assuming a two-tailed test) was chosen as the level of statistical significance in this study. This analytical approach established each subject as his own control and thereby reduced the effects of biological variability among subjects.

Evaluation of the entire measured acceleration-time histories of head and chest was also accomplished by calculated severity indices (2). These single parameters, which were derived by a weighted integral of the acceleration-time function taken over the interval of the impact ($SI = \int a^n(t)dt$, where $n = 2.5$), were used to compare the overall severities of impact responses. No significance was assigned to the absolute values of these measures. Instead, they were utilized only in a relative sense for comparison.

EXPERIMENTAL RESULTS

Data from 100 selected impact tests (54 at the 8 G level and 46 at the 10 G level) conducted between January and August 1981 are presented. The three preload conditions allowed these to be sorted into 92 comparable test pairs which were matched for velocity change of the sled. The difference in velocity change for any given test pair was $\leq$ 1 ft/sec. (The velocity difference in 88% of the 92 matched pairs was $\leq$ 0.5 ft/sec.) Nineteen subjects (17 males and 2 females) participated in the test program and twelve subjects completed all test conditions. Four subjects did not complete the test program since they departed Wright-Patterson Air Force Base for new duty assignments; one subject was temporarily medically disqualified from participation; and two subjects voluntarily withdrew from the program.

Tests on the Horizontal Decelerator were conducted first. The phases of an experimental run on this facility are shown in Figure 3. In these tests, the sled was accelerated to a velocity sufficiently high to achieve the programmed impact velocity at the decelerator following the deceleration experienced during the coast phase. During the velocity control window of the coast phase, the actual sled velocity was compared to a programmed or model velocity. If the actual velocity was higher than programmed, the excess velocity was "trimmed" by the application of sled-mounted braking devices. In this way, the actual sled velocity at impact was assured to be within 1 ft/sec of the desired impact velocity. In Test Conditions A and D, the trim brakes were applied throughout the 250 msec just prior to impact in order to impose an additional approximate 0.35 G dynamic preload on the sled. To achieve the desired impact velocity during these test conditions, it was necessary to allow for the higher velocity change associated with the additional imposed dynamic preload. The sled was, therefore, programmed to leave the velocity control window of the coast phase with a higher actual velocity than in Test Conditions B and E. The 8 G tests were conducted prior to the 10 G tests on the decelerator. Subjects were informed of the impact test level but were not informed of the preload condition prior to each test. The order of presentation of the preload conditions was randomized.

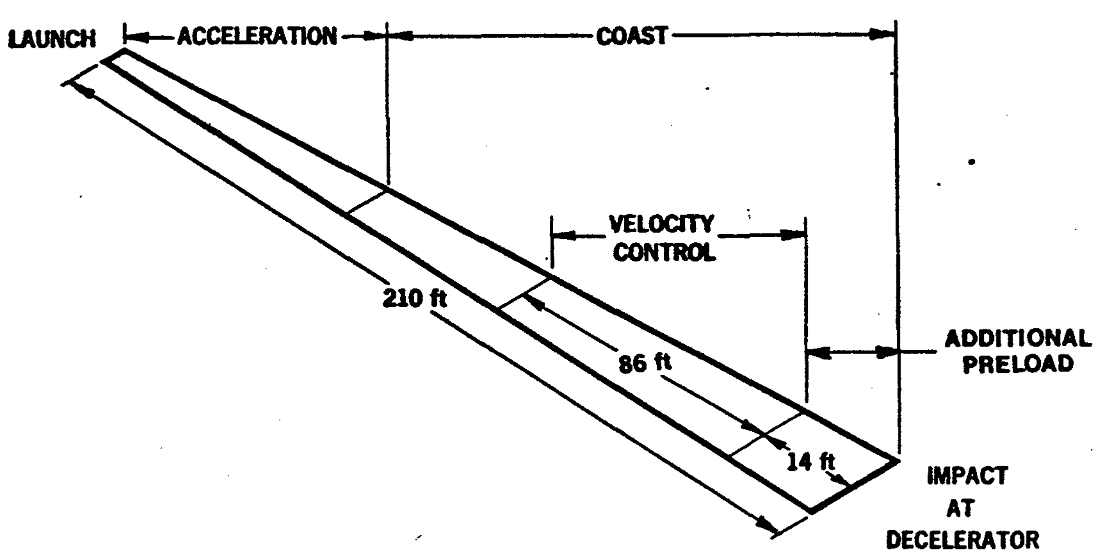

Figure 3. Phases of the Experimental Run on the Horizontal Decelerator

Tests on the Impulse Accelerator were conducted following completion of all decelerator tests. These tests were not randomized with the deceleration conditions, since transitioning from decelerator to accelerator required substantial facility modifications. However, the difference in mechanization would have prevented blind or double blind presentation anyway. The accelerator tests were designed to produce an impact velocity as close as possible to that measured for the same subject on the decelerator. Since the profiles were almost identical (see Figures 1 and 2), velocity matching implied close correspondence in acceleration as well. However, the mean sled acceleration peak for the 8 G tests was higher on the accelerator (Test Condition C) than for either comparable decelerator condition (Test Conditions A and B). These small differences in the mean sled acceleration peak (0.18 G in comparison B-C and 0.26 G in comparison A-C) were systematic enough to appear as statistically significant in the Wilcoxon analyses. (See tables in Appendix.) To assure that this test bias was not the basis for the observed differences in subject response, 10 G tests on the accelerator were designed to produce slightly lower velocity changes for each subject than had been observed on the decelerator. Sled acceleration peaks for the 10 G tests were, therefore, lower on the accelerator (Test Condition F) than for either comparable decelerator condition (Test Conditions D and E). These small differences (0.36 G in comparison E-F and 0.30 G in comparison D-F) were again systematic enough to appear as statistically significant in the Wilcoxon analyses, but this time, by design, in the other direction. Despite this variation in sled acceleration, response differences at 10 G similar to those documented at 8 G were still observed.

Six comparisons among the test conditions were made. The means and standard deviations of the measured parameters and the statistically significant trends established by the Wilcoxon analysis at the 90% confidence level are presented for each comparison in Tables A1-A6. (See Appendix.) In these tables, an asterisk (*) indicates a statistically significant change in the designated parameter. Tables 4 and 5 summarize the statistically significant trends in all comparisons of the electronic data. In these tables, the arrow designates the direction of the trend, the number indicates the percentage increase in the parameter mean between test conditions, and an asterisk again designates statistical significance at the chosen confidence level.

TABLE 4
SUMMARY OF STATISTICALLY SIGNIFICANT TRENDS
AND PERCENT INCREASE IN PARAMETER MEANS
FOR COMPARISONS AT THE 8 G TEST LEVEL

TEST CONDITIONS	A / B	B / C	A / C
TEST FACILITY	Dec / Dec	Dec / Acc	Dec / Acc
DYNAMIC PRELOAD (G)	0.62 / 0.25	0.25 / 0	0.62 / 0
SM SLED ACCELERATION	---> 1	* ---> 2	* ---> 3
SLED VELOCITY	1 <--- *	0	1 <---
CHEST ACCELERATION			
-X axis	---> 3	---> 4	* ---> 7
+Z axis	* ---> 11	* ---> 42	* ---> 53
Resultant	---> 3	* ---> 16	* ---> 18
CHEST SEVERITY INDEX	* ---> 12	* ---> 25	* ---> 38
HEAD ACCELERATION			
-X axis	---> 5	* ---> 36	* ---> 40
-Z axis	* ---> 25	* ---> 119	* ---> 170
Resultant	* ---> 9	* ---> 47	* ---> 58
HEAD SEVERITY INDEX	---> 15	* ---> 88	* ---> 113
STRAP LOADS			
Total Shoulder Straps	---> 3	* ---> 8	* ---> 10
Total Lap Belt	* ---> 3	* ---> 13	* ---> 15
SEAT PAN LOADS			
+Z axis	* ---> 5	* ---> 8	* ---> 11
Resultant	* ---> 4	* ---> 6	* ---> 9

Comparisons of the 8 G test results revealed statistically significant increases in measured and computed response parameters in the test condition with less dynamic preload. (See Table 4.) In comparison A-B, for example, statistically significant increases in resultant head acceleration, total lap belt load, and resultant seat pan load were seen in the condition with less dynamic preload (Test Condition B). Increases in the same direction were seen for vertical chest acceleration and the chest Severity Index. More dramatic changes in response parameters were seen in the decelerator-accelerator comparisons. In comparison B-C the resultant chest acceleration and chest Severity Index, resultant head acceleration and head Severity Index, total shoulder strap load, total lap belt load, and resultant seat pan load were increased in the exposure on the accelerator. Similar changes, all larger in magnitude, were seen in the A-C comparison. The direction of these trends and the relative magnitudes of the increases observed in this comparison may have been anticipated on the basis of the findings in the A-B and B-C comparisons, since the three comparisons are not independent.

Comparisons of the 10 G test results also indicated statistically significant increases in the measured and computed parameters in the test condition with less dynamic preload. For example, in the D-E comparison, statistically significant increases are seen

in resultant chest acceleration and chest Severity Index in the condition with less dynamic preload (Test Condition E). Although similar findings at the chest are absent in the decelerator-accelerator comparisons, the resultant head acceleration and head Severity Index, total lap belt load, and vertical seat pan reaction load were increased in Test Condition F relative to either comparable test condition on the decelerator (Test Conditions E and D). These findings are summarized in Table 5.

TABLE 5
SUMMARY OF STATISTICALLY SIGNIFICANT TRENDS
AND PERCENT INCREASE IN PARAMETER MEANS
FOR COMPARISONS AT THE 10 G TEST LEVEL

TEST CONDITIONS	D	E	E	F	D	F
TEST FACILITY	Dec	Dec	Dec	Acc	Dec	Acc
DYNAMIC PRELOAD (G)	0.62	0.26	0.26	0	0.62	0
SM SLED ACCELERATION	0		4 <---	*	3 <---	*
SLED VELOCITY	0		0		0	
CHEST ACCELERATION						
-X axis	* --->	3	4 <---		--->	2
+Z axis	* --->	12	* --->	38	* --->	47
Resultant	* --->	4	--->	7	--->	9
CHEST SEVERITY INDEX	* --->	9	--->	7	--->	14
HEAD ACCELERATION						
-X axis	0		* --->	25	* --->	22
-Z axis	--->	11	* --->	61	* --->	88
Resultant	--->	4	* --->	31	* --->	36
HEAD SEVERITY INDEX	--->	5	* --->	56	* --->	62
STRAP LOADS						
Total Shoulder Straps	--->	2	2 <---		0	
Total Lap Belt	--->	1	* --->	6	* --->	8
SEAT PAN LOADS						
+Z axis	--->	2	* --->	6	* --->	8
Resultant	--->	2	--->	5	* --->	8

These electronic data indicate that forces and body segment accelerations imposed on subjects in impacts preceded even by a minimal dynamic preload are decreased in comparison to those measured during impacts preceded by less dynamic preload. These changes were more dramatic in the decelerator-accelerator comparisons than in the comparisons involving the two decelerator preload conditions. This was true despite the fact that, in the latter comparisons, the difference in dynamic preload between the two decelerator test conditions (approximately 0.35 G) was greater than the difference in dynamic preload between the low preload decelerator case and the no preload accelerator case (approximately 0.25 G). Furthermore, the threshold for significant dynamic preload effects on impact response occurs below 0.25 G. Determining the minimum threshold for these effects would require experiments on a decelerator facility which would impose less than 0.25 G dynamic preload on the test vehicle.

The differences in the statistically significant trends at the 8 G test level (Table 4) relative to those at the 10 G level (Table 5) are probably attributable to variations in test conditions. The variations in sled acceleration peak have already been described. In addition, the pulse duration of the 8 G exposure was different from that of the 10 G exposure, implying differences in the frequency content of these impact waveforms. The mechanical response of the subjects at the two test levels, therefore, should be dissimilar and may account for the differences in observed preload effect.

TABLE 6
SUMMARY OF OBJECTIVE MEDICAL FINDINGS

TEST LEVEL (G)	8	8	8	10	10	10
TEST CONDITION	A	B	C	D	E	F†
DYNAMIC PRELOAD (G)	0.62	0.25	0	0.62	0.26	0
n =	19	19	16	17	17	12
ABRASIONS	1	1	8	1	2	5
CONTUSIONS	0	0	3	0	2	2
MUSCLE STRAINS	0	1	2	0	0	2

† Two subjects declined this 10 G exposure. Two subjects who participated utilized adhesive tape at the clavicles to prevent abrasions during this 10 G exposure only.

The adverse medical effects of subject participation were confined to anticipated and clinically inconsequential abrasions, contusions, and muscle strains (with the exception of a possible minimal Type I atlantoaxial rotatory fixation in one subject). These data are summarized in Table 6. The abrasions, of course, are observed in areas of subject contact with the restraint straps, particularly at the clavicles, and are easily identifiable post-impact. However, contusions and muscle strains have probably been underestimated, since these effects may not be seen immediately following the impact and may simply not be reported by subjects in follow-up. Nevertheless, the frequency of

these objective medical findings following impact exposures on the accelerator was increased compared to similar exposures on the decelerator at both the 8 G and 10 G test levels. However, the frequency of adverse effects incurred on the accelerator did not increase from the 8 G to the 10 G test level. No attempt was made to differentiate levels of injury severity. The relative scarcity of medical findings in the decelerator tests suggests an effective decrease in the threshold for abrasions with imposed dynamic preload. At higher impact acceleration levels, such as those experienced operationally during aircraft ejection, it is conceivable that a similar beneficial threshold shift may occur, with preload, for the clinically consequential adverse effects of these impacts, such as vertebral fractures.

Analysis of the subject questionnaire responses indicated that subjects, in general, perceived their impact response on the accelerator to be more severe than their response to comparable decelerator tests. For example, in the B-C comparison, 9 of 16 subjects indicated that the overall impact was more comfortable on the decelerator than on the accelerator. Five subjects indicated no difference between the two test conditions and two subjects indicated that the overall response was more comfortable on the accelerator than on the decelerator. Similarly, at the 10 G test level, in comparison E-F, 8 of 12 subjects indicated their overall response on the decelerator was more comfortable than on the accelerator, three subjects indicated no difference and one subject indicated that the overall response on the accelerator was less severe than on the decelerator. The numerical average of the subject responses to this question were computed for these test conditions. These averages indicated the consensus of the subjects that the overall severity of a 10 G impact exposure preceded by nominal track friction (0.25 G) on the decelerator was equivalent to the overall severity of an 8 G impact exposure on the accelerator. In fact, the two subjects who voluntarily withdrew from the test program declined the 10 G accelerator exposure following completion of the 8 G exposure on the accelerator and at least one 10 G test on the decelerator. Interestingly enough, comparison of objective measures of accelerations and forces between 10 G decelerator tests and 8 G accelerator tests for the same subjects supports these subjective assessments. Resultant head acceleration was higher in the 8 G accelerator tests, while resultant chest acceleration, harness loads, and seat forces were higher in the 10 G decelerator tests.

DISCUSSION

The test results demonstrate statistically significant increases in severity of human impact response when compared to response measured in similar impacts preceded by higher dynamic preload. These changes are particularly striking when accelerator impacts are compared to matched decelerator impacts. These response differences continue to be statistically significant, in most cases, even when the accelerator event is less severe, as seen in the 10 G comparisons E-F and D-F.

The explanation for the response differences seen in this test program can be understood in part by examining the concept of dead space and the nature of viscoelastic systems exposed to impact. In spite of pretensioning, some structures of the human body are poorly supported by the restraint system. In typical systems, these include the head, arms, legs, and various soft tissue and internal organs. These structures often must displace before direct accelerating forces can be applied through structural attachments. This amounts to a functional "dead space", which effectively delays the onset of acceleration and implies an eventual increased magnitude of acceleration to allow the late-starting member to "catch up". A dead space mechanism such as this is more observable externally in tests with high dynamic preload, such as some reported by Stapp, in which the head and extremities are actually thrown forward during application of preload. In the tests reported here, the dead space mechanism would be less observable and of lower magnitude, but still may occur internally.

The initial conditions imposed on the viscoelastic system of subject, support, and restraint are modified by dynamic preloading. The initial conditions in the decelerator tests are observably different from those in accelerator tests. In the former, loads in harnesses and forces on the seat structure changed during the transition from launch to coast. At impact, therefore, the viscoelastic response of the subject had already begun. All portions of the subject respond to impact partially as springs, and these springs had already begun to deform while under dynamic preload. Such anticipatory deformation has an effect similar to that of removing simple dead space in the sense that the subject response can follow the acceleration of the supporting structure more closely. However, unlike simple dead space, which has no spring constant, viscoelastic deformation of the entire structure must take place under dynamic loading which produces whole body acceleration. The overall effect simply cannot be duplicated by static pretensioning of harness systems or by voluntary bracing.

The apparent protective effects of dynamic preload have two significant implications. The first is that our assumptions about human tolerance should be re-examined. The ability of a human to tolerate a high-energy 45 G impact with significant dynamic preload does not imply that similarly capable subjects can tolerate a similar impact without preload. The acceleration-time history prior to the impact must be specified and scaling laws must be devised in order to improve the comparability of tests conducted on different impact facilities. The second implication of these results is more positive. If dynamic preload makes impact more tolerable, it should be exploitable in impact protection systems.

Practical utilization of dynamic preload requires that the coming impact be sensed in time to allow application of the preload. During this program, in Test Conditions B and E, dynamic preload was applied over a period of approximately 3 seconds of coasting and, in Test Conditions A and D, an additional approximate 0.35 G dynamic preload was applied during the 250 msec immediately prior to the impact event. For practical impact protection, shorter durations of preload application may be required, as well as a means to impose the preload in coordination with the impact. Unplanned impacts, such as crashes, would require impact initiation sensors at the vehicle periphery or beyond it. Planned impacts, such as ejection seat firing, could use preload during the pre-ejection sequence. For either case, the minimum duration of a protective preload pulse must still be determined. The optimum magnitude and duration will depend upon the dynamic mechanical response properties of the subject and restraint system and the characteristics of the impact to be experienced. Furthermore, the direction of the preloading force should be along the impact force vector.

Work at AFAMRL is continuing in order to define practical applications of dynamic preload for use in aircraft escape systems. This application is particularly attractive, since idealized preloading pulses have also been shown to promise improvement in the displacement-time performance of the seat (6). Thus, it may be possible to improve the performance of the seat, and thus its envelope, while at the same time imposing a protective dynamic preload on the seat occupant in order to decrease the probability of injury. The potential for practical and realizable escape systems incorporating dynamic preload will be defined by measuring human response to various characteristic preloading waveforms in vertical impact.

REFERENCES

1. Beeding, E. L., Daisy Decelerator Tests (13 July 1959 - 13 April 1960), MDW Test Report No. 60-4, Air Force Missile Development Center, Holloman Air Force Base, New Mexico, July 1960.

2. Gadd, C. W., "Use of a Weighted-Impulse Criterion for Estimating Injury Hazard" in Tenth Stapp Car Crash Conference, Society of Automotive Engineers, Inc., New York, November 1966.

3. Hearon, B. F. and J. H. Raddin, Jr., "Experience with Highly Selective Screening Techniques for Acceleration Stress Duty" in The Effect of Long-Term Therapeutics, Prophylaxis and Screening Techniques on Aircrew Medical Standards, AGARD-CP-310, March 1981.

4. Lombard, C. F., Collected Data on 48 Rocket Sled Experiments (Holloman AFB), NSL 65-94, Northrop Space Laboratories, 1965.

5. Majewski, P. L., T. J. Borgman, D. J. Thomas, and C. L. Ewing, "Transient Intraventricular Conduction Defects Observed During Experimental Impacts in Human Subjects" in Models and Analogues for the Evaluation of Human Biodynamic Response, Performance and Protection, AGARD-CPP-253, November 1978.

6. Payne, P. R. and D. A. Shaffer, An Optimum Acceleration-Time History for an Escape System, AMRL-TR-70-143, Aerospace Medical Research Laboratory, Wright-Patterson Air Force Base, Ohio, 1971.

7. Raddin, J. H., Jr., J. W. Brinkley, and B. F. Hearon, "The Implications of Dynamic Preload in Human Impact Tolerance" presented at the 51st Annual Scientific Meeting of the Aerospace Medical Association (Anaheim, California), May 1980.

8. Stapp, J. P., Human Exposures to Linear Deceleration: Part 2 - The Forward-Facing Position and the Development of a Crash Harness, AF-TR-5915, Part 2, December 1951.

9. Wilcoxon, F. and R. A. Wilcox, Some Rapid Approximate Statistical Procedures, Lederle Laboratories, New York, 1964.

APPENDIX

TABLE A1

COMPARISON A-B (8 G)
SUMMARY OF ELECTRONICALLY MEASURED AND COMPUTED DATA FROM WILCOXON ANALYSIS
(Peak values are tabulated for velocity, accelerations and loads.)
(n = 19)

TEST CONDITION TEST FACILITY DYNAMIC PRELOAD (G)	A Decelerator 0.62		B Decelerator 0.25		Significant at 90% Confidence
	Mean	SD	Mean	SD	
SM SLED ACCELERATION (G)	8.17	0.20	8.26	0.28	
SLED VELOCITY (ft/sec)	27.9	0.31	27.7	0.45	*
CHEST ACCELERATION (G)					
-X axis	-9.48	1.41	-9.77	0.91	
+Z axis	5.77	1.31	6.40	1.23	*
Resultant	10.6	0.89	10.9	0.74	
CHEST SEVERITY INDEX	18.6	1.92	20.9	2.65	*
HEAD ACCELERATION (G)					
-X axis	-9.86	1.78	-10.4	2.19	
-Z axis	-4.11	1.75	-5.12	2.79	*
Resultant	10.6	1.96	11.6	2.78	*
HEAD SEVERITY INDEX	24.1	6.69	27.8	10.5	
STRAP LOADS (lb)					
Total Shoulder Straps	519	103	534	87	
Total Lap Belt	1230	191	1270	179	*
SEAT PAN LOADS (lb)					
+Z axis	1070	214	1120	215	*
Resultant	1130	210	1180	213	*

TABLE A2

COMPARISON B-C (8 G)
SUMMARY OF ELECTRONICALLY MEASURED AND COMPUTED DATA FROM WILCOXON ANALYSIS
(Peak values are tabulated for velocity, accelerations and loads.)
(n = 16)

TEST CONDITION TEST FACILITY DYNAMIC PRELOAD (G)	B Decelerator 0.25		C Accelerator 0		Significant at 90% Confidence
	Mean	SD	Mean	SD	
SM SLED ACCELERATION (G)	8.23	0.23	8.41	0.16	*
SLED VELOCITY (ft/sec)	27.6	0.36	27.6	0.41	
CHEST ACCELERATION (G)					
-X axis	-9.58	0.79	-9.98	1.23	
+Z axis	6.47	1.09	9.17	4.34	*
Resultant	10.8	0.70	12.5	2.97	*
CHEST SEVERITY INDEX	20.8	2.70	26.0	6.53	*
HEAD ACCELERATION (G)					
-X axis	-10.3	2.29	-14.0	3.29	*
-Z axis	-4.85	2.59	-10.6	2.57	*
Resultant	11.4	2.82	16.8	3.31	*
HEAD SEVERITY INDEX	27.0	9.97	50.8	16.0	*
STRAP LOADS (lb)					
Total Shoulder Straps	535	86	577	102	*
Total Lap Belt	1280	183	1440	239	*
SEAT PAN LOADS (lb)					
+Z axis	1130	224	1220	236	*
Resultant	1190	225	1260	235	*

TABLE A3

COMPARISON A-C (8 G)
SUMMARY OF ELECTRONICALLY MEASURED AND COMPUTED DATA FROM WILCOXON ANALYSIS
(Peak values are tabulated for velocity, accelerations and loads.)
(n = 16)

TEST CONDITION TEST FACILITY DYNAMIC PRELOAD (G)	A Decelerator 0.62		C Accelerator 0		Significant at 90% Confidence
	Mean	SD	Mean	SD	
SM SLED ACCELERATION (G)	8.15	0.21	8.41	0.16	*
SLED VELOCITY (ft/sec)	27.8	0.32	27.6	0.41	
CHEST ACCELERATION (G)					
-X axis	-9.36	1.42	-9.98	1.23	*
+Z axis	5.99	1.30	9.17	4.34	*
Resultant	10.6	0.84	12.5	2.97	*
CHEST SEVERITY INDEX	18.8	1.91	26.0	6.53	*
HEAD ACCELERATION (G)					
-X axis	-9.97	1.87	-14.0	3.29	*
-Z axis	-3.93	1.65	-10.6	2.57	*
Resultant	10.6	2.04	16.8	3.31	*
HEAD SEVERITY INDEX	23.8	6.77	50.8	16.0	*
STRAP LOADS (lb)					
Total Shoulder Straps	526	101	577	102	*
Total Lap Belt	1250	171	1440	239	*
SEAT PAN LOADS (lb)					
+Z axis	1100	215	1220	236	*
Resultant	1160	216	1260	235	*

TABLE A4

COMPARISON D-E (10 G)
SUMMARY OF ELECTRONICALLY MEASURED AND COMPUTED DATA FROM WILCOXON ANALYSIS
(Peak values are tabulated for velocity, accelerations and loads.)
(n = 17)

TEST CONDITION TEST FACILITY DYNAMIC PRELOAD (G)	D Decelerator 0.62		E Decelerator 0.26		Significant at 90% Confidence
	Mean	SD	Mean	SD	
SM SLED ACCELERATION (G)	9.82	0.19	9.85	0.19	
SLED VELOCITY (ft/sec)	30.5	0.29	30.4	0.32	
CHEST ACCELERATION (G)					
-X axis	-11.7	1.14	-12.1	1.31	*
+Z axis †	6.51	1.12	7.31	1.33	*
Resultant †	12.8	0.91	13.3	0.87	*
CHEST SEVERITY INDEX †	28.1	2.72	30.6	3.40	*
HEAD ACCELERATION (G)					
-X axis	-12.3	2.00	-12.3	2.52	
-Z axis	-6.62	3.25	-7.36	4.37	
Resultant	13.8	3.08	14.3	4.20	
HEAD SEVERITY INDEX	40.0	14.4	42.1	18.5	
STRAP LOADS (lb)					
Total Shoulder Straps	666	110	677	113	
Total Lap Belt †	1530	216	1550	219	
SEAT PAN LOADS (lb)					
+Z axis	1290	236	1320	238	
Resultant	1350	235	1380	233	

(† These parameters based on n = 14 due to partial data loss in three tests.)

TABLE A5

COMPARISON E-F (10 G)
SUMMARY OF ELECTRONICALLY MEASURED AND COMPUTED DATA FROM WILCOXON ANALYSIS
(Peak values are tabulated for velocity, accelerations and loads.)
(n = 12)

TEST CONDITION TEST FACILITY DYNAMIC PRELOAD (G)	E Decelerator 0.26		F Accelerator 0		Significant at 90% Confidence
	Mean	SD	Mean	SD	
SM SLED ACCELERATION (G)	9.91	0.19	9.55	0.22	*
SLED VELOCITY (ft/sec)	30.5	0.34	30.4	0.50	
CHEST ACCELERATION (G)					
-X axis	-12.2	1.32	-11.7	1.51	
+Z axis	7.46	0.91	10.3	4.17	*
Resultant	13.3	0.96	14.2	2.70	
CHEST SEVERITY INDEX	31.2	3.33	33.5	6.33	
HEAD ACCELERATION (G)					
-X axis	-12.2	2.42	-15.0	2.70	*
-Z axis	-7.15	3.67	-11.5	2.72	*
Resultant	14.0	3.76	18.4	3.32	*
HEAD SEVERITY INDEX	41.6	17.4	64.8	17.6	*
STRAP LOADS (lb)					
Total Shoulder Straps	676	116	665	99	
Total Lap Belt	1590	167	1680	246	*
SEAT PAN LOADS (lb)					
+Z axis	1330	198	1410	193	*
Resultant	1390	201	1460	193	

TABLE A6

COMPARISON D-F (10 G)
SUMMARY OF ELECTRONICALLY MEASURED AND COMPUTED DATA FROM WILCOXON ANALYSIS
(Peak values are tabulated for velocity, accelerations and loads.)
(n = 12)

TEST CONDITION TEST FACILITY DYNAMIC PRELOAD (G)	D Decelerator 0.62		F Accelerator 0		Significant at 90% Confidence
	Mean	SD	Mean	SD	
SM SLED ACCELERATION (G)	9.85	0.21	9.55	0.22	*
SLED VELOCITY (ft/sec)	30.5	0.32	30.4	0.50	
CHEST ACCELERATION (G)					
-X axis	-11.5	1.19	-11.7	1.51	
+Z axis †	6.79	0.78	10.0	4.50	*
Resultant †	12.8	1.06	14.0	2.92	
CHEST SEVERITY INDEX †	28.7	3.03	32.8	6.74	
HEAD ACCELERATION (G)					
-X axis	-12.3	1.94	-15.0	2.70	*
-Z axis	-6.12	2.39	-11.5	2.72	*
Resultant	13.5	2.38	18.4	3.32	*
HEAD SEVERITY INDEX	39.9	13.5	64.8	17.6	*
STRAP LOADS (lb)					
Total Shoulder Straps	668	108	665	99	
Total Lap Belt †	1550	190	1680	241	*
SEAT PAN LOADS (lb)					
+Z axis	1300	182	1410	193	*
Resultant	1350	185	1460	193	*

(† These parameters based on n = 10 due to partial data loss in two tests.)

DISCUSSION

S. P. DESJARDINS (US)

Comment: Your findings are of special interest in our understanding and influencing energy-absorbing seat performance. The new crashworthy US Army helicopters include energy-absorbing landing gear which provides the preload and thus perhaps the benefits you describe in your paper. In our qualification drop test of the AH-64A crewseat we simulated the energy-absorbing stroke of the gear in our test pulse. The seat performance was superior with fewer high amplitude transients than measured on other seats where the standard triangular test pulse was used.

AUTHOR

In the helicopter described, the downward, passive stroking landing gear will modify the vertical acceleration measured at the seat. If this modification takes the form of $+G_z$ acceleration preceding and continuous with the $+G_z$ impact event, then dynamic preload, by our definition, will have been applied to the seat and occupant. The effects of such anticipatory vertical acceleration on human impact response will be investigated in a future test program at AFAMRL.

However, a distinction must be drawn between downward, passive stroking or energy-absorbing seats and upward, active stroking mechanisms by which vertical dynamic preload conceivably may be applied to a seat occupant. The former appear to impart a beneficial effect during the impact event by limiting the imposed acceleration. On the other hand, the latter would impose an acceleration before an anticipated impact and, in so doing, better prepare the seat occupant viscoelastically for the event. (It may also be conceivable that an upward, active stroke could be utilized in conjunction with a later downward, passive stroke to combine the beneficial effects of each protection technique.) The timing, amplitude, and duration of this dynamic preload pulse would be critical in order to have a beneficial effect. The effects of variations in these preload parameters are the subject of a current AFAMRL human test program.

DR. D. J. THOMAS (US)

1. What was the head and neck initial condition variability within subjects between runs?

2. What was the angular acceleration of the head for each run and was this accounted for in the statistical analysis of the acceleration peaks measured at the mouth?

3. What is unique about the preloading effect that cannot be explained by variation due to initial condition, head angular acceleration resopnse, and forceful loading of the restraint system, all of which are well described from prior experiments?

AUTHOR

1. The test seat and restraint geometry was not varied during this test program. The vertical position of the headrest was varied among subjects, but was the same for a given subject in all test conditions. To further minimize variations in the head-neck initial conditon prior to each experiment, subjects were instructed to assume the same body position they had assumed in previous tests. In particular, each subject was asked to keep his head back against the headrest (head up, chin up) and to maintain a mild to moderate amount of neck muscle tension. Proper head position was verified by the test conductor prior to each experiment.

In addition, the effects of variations in initial conditions among subjects on data analysis were minimized by use of the Wilcoxon paired-replicate rank test to establish the statistical significance of results. In this technique, the test results of each subject are compared only to that subject's results in other test conditions. Thus, each subject is his own control, minimizing the influence of biological variability and small differences in initial positions among subjects on data analysis.

2. Direct measurements of angular head accelerations were not made in this study. A triaxial translational accelerometer package was used to obtain the head (and chest) acceleration data. This device, of course, measured translational acceleration components summed with translational components resulting from angular motions. The rotational motion of the head was measured photometrically. These data are being analyzed to derive head angular velocities and accelerations.

3. The static initial conditions in the present study were very carefully controlled to minimize variations in, e.g., subject position and bracing as well as static pretensioning of the restraint. In addition, the impact event profiles at comparable test levels were nearly identical. The observed response differences in the test conditions must, therefore, be attributed to the differences in the pre-event acceleration time history.

Dynamic preload may be distinguished from other impact protection techniques in that it does not require active participation of the subject and, more importantly, in that it involves a whole body viscoelastic preparation of the subject for the impact event. The potential aerospace applications of dynamic preload also set it apart from other protection techniques. If vertical preload pulses can be demonstrated to ammeloriate human response to subsequent vertical impact, then utilizing dynamic preload prior to current ejection seat profiles would be expected to reduce crewmember morbidity during emergency escape. At the same time, such an impulsive velocity change will improve the displacement-time performance of the ejection seat. A single modification to the acceleration profile could, therefore, conceivably have two separate, beneficial effects. Similar effects are not seen with other protection techniques.